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Influence of Constraining Barrier on the 5th Lumbar and 1st Sacral joint Compressive Force during Manual Lifting

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Abstract. The purpose of this study was to examine the mechanical stresses on the lower back as the response of different heights of constraining barrier. Ten male subjects lifted a load from the floor to the knuckle height under the non-constrained and the constrained conditions with 4 different heights of constraining barrier (80%, 100%, 120% and 140% of knee height). The constrained condition was defined as the condition where a load was placed on the floor behind a certain level of bar. When lifting of the constrained conditions, subjects significantly increased the peak compressive forces at L5/S1 compared to the non-constrained (3868.8 ± 527.5 N, 4175.0 ± 486.0 N, 4162.4 ± 462.3 N, 4136.0 ± 553.1 N, 4079.4 ± 468.9 N for 0%, 80%, 100%, 120% and 140% barrier height conditions respectively). The subjects moved the load further from L5/S1 in the horizontal direction when lifting during the constrained conditions. While lifting during the constrained conditions subjects generated an increase in the sacral angle and a decrease of the knee flexion. The peak compressive forces at L5/S1 showed a statistically significant quadratic trend. However, the magnitude of the difference of peak compressive forces during the constrained conditions was small.

Keywords: *back injury; constraint barrier; L5/S1 disc; manual lifting; peak compressive forces.*

1 Introduction

Epidemiologic research indicates that lower back pain is a major problem in terms of both humans suffering and cost for workers. Manual material handling tasks, especially repetitive lifting, is most commonly reported as the cause of back injuries (Chaffin and Park, 1973, Frymoyer et al., 1983, Klein et al., 1984). This is because the repetitive lifting produces high compressive stresses on the back, especially the lower back (Dolan et al., 1994), which can cause the

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degeneration on the annulus fibrosus of the intervertebral discs and then cause the lumbar disc to prolapse posteriorly (Adams and Hutton, 1983, 1985).

Researchers have developed a considerable interest in how people apply the motion and strategy of lifting and how people control the effect of lifting on the human body (NIOSH, 1981). These studies that have focused on the stress occurring in the lower back have utilized a biomechanical model (De Looze, et al., 1992, Schipplein et al., 1990, Tsuang et al., 1992). In this approach, the compressive forces acting on the lower back are estimated based on the reactive forces at the L5/S1 intervertebral disc center, knowledge of musculature and the sacral orientation. The models assume rigid body segments connected by hinge joints. The magnitude of loads being lifted and anthropometric measurements of body segments are required along with knowledge of external forces acting on the body. Applying conventional Newtonian equations of motion, the joint reactive forces and moments are predicted and L5/S1 compressive forces are calculated.

Individuals sometimes require lifting from, and lowering into, industrial bins. These bins form a constraint barrier that can serve to restrict the preferred motion of the body and result in altered stresses. Lifting tasks with different barrier heights can influence lifting kinematics, moments, and compressive forces at L5/S1 (the joint between the fifth lumbar and the first sacral vertebrae) during manual lifting. McKean and Potvin (2001) studied the effects of a simulated industrial bin on lifting and lowering posture and trunk muscle activity (EMG). The wall height of the simulated industrial bin was constructed at 120% of the average male/female knee height. Male subjects lifted a 15 kg load and female subjects lifted an 8.5 kg load. Each subject performed 10 lifts and 10 lowers under both "freestyle" and "constrained" conditions. The load was generally lifted and lowered at a greater peak horizontal displacement for the constrained condition than the freestyle condition. Subjects tended to have larger peak pelvis and trunk flexion angles, but less peak knee angle for the constrained condition than for the freestyle condition. The peak EMG magnitudes for both thoracic and lumbar muscle groups also were higher for the constrained than for the freestyle condition.

The study of McKean and Potvin (2001) was one of the only studies to look at the effects of a constraining barrier on lifting and lowering, even though it has implications to many material handling tasks applied in industry. However, this study did not directly calculate the mechanical stresses (i.e., moments, compressive forces) that occurred on the lower back in response to a constraining barrier. They limited their study to the changes in the magnitude of back muscle activity (EMG) to represent the loading of the back. Although the EMG is related to the muscle forces in the lower back, it is an indirect measure and and it is only one of the 2 components that make up the compressive force. The other component is the reaction force, which was measured in this study using a rigid body model, external reaction forces and inverse dynamics procedures.

Another limitation of the previous study (McKean & Potvin, 2001) was the measurement of a single simulated bin height (120%). Then, in this study, by adding additional bin heights it is possible to determine if there is a linear change in compressive forces as bin height increases or if there is a threshold bin height that should be avoided.

The purpose of this study was to examine the mechanical stresses (compressive force) on the lower back in response to the different levels of constraining barrier. It was hypothesized that lifting tasks over a constraining barrier would generate greater stresses on the lower back than lifting tasks over a non-constraining barrier. This was also hypothesized that the higher heights of constraining barrier, the greater stresses on the lower back would be generated. The heights of the constraining barrier were constructed to be 0%, 80%, 100%, 120% and 140% of knee height.

2 Methods

2.1 Subjects

Ten healthy young male subjects with no history of back injuries participated in this study. The mean age was 25 ± 4.9 years; body mass was 75 ± 8.8 kg; and body height was 1.75 ± 0.06 m. The subjects signed informed consent to participate in this study in accordance with university policy. Prior to the start of the study, subjects were familiarized with the experiment protocol.

2.2 Protocol

Each subject was asked to lift a load under 5 different conditions. Each condition consisted of a series of 10 lifts. The conditions were the non-constrained (as a 'free-normal' lifting) condition and the constrained conditions, with 4 different heights of constraining barrier. The constrained condition was defined as the condition where the load was placed on the floor behind a certain height of constraining barrier. The constraining barrier was a bar that placed between the subject and the load. The heights of the constraining barrier were constructed to be 0%, 80%, 100%, 120% and 140% of the average male knee height. The knee height was calculated as 28.5% of total body height (Chaffin and Andersson, 1999). The order of presentation of these conditions was balanced for each subject. During each condition a lift was performed every 6 seconds based on an audible tone. The subjects were permitted as much as rest as needed between conditions.

The load was a crate (42 cm length, 34.5 cm width, 27 cm height) that contained a mass of 10.3 kg placed on the floor in front of the subject. The crate had two fixed handles placed symmetrically 27 cm above the bottom. The handles and mass center of the crate were positioned approximately 27 cm horizontally from the subject's ankles.

Each subject wore shoes during the experiment. The right foot of each subject was placed on the force platform during the experiment. Subjects began the lifting experiment from the standing position, then were asked to bend down, grab the crate and returned to a standing position as they lifted the crate. The crate was then returned to the original position. Subjects were encouraged to perform lifts at a 'normal' speed and use a natural lifting technique.

3 Model

The motions in this study were recorded using four 120 Hz video cameras (Peak Performance Technologies, Inc., 1997). The three-dimensional positions of nine reflective markers were recorded (figure 1). Reflective markers were placed on the second toe, the posterior point of the heel, the lateral malleolus, the midpoint of the lateral joint line of the knee, and the glenohumeral joint. Three additional markers were then placed on the pelvis region to help defining the joint between the fifth lumbar and first sacral vertebrae (L5/S1). These markers were located at the end of an 8-cm wand placed over the anterior iliac crest and over the posterior sacrum. These markers were used to form a local coordinate system with the hip marker as the origin. A final marker was placed on the crate. The three dimensional coordinates were automatically calculated and were then low-pass filtered with a fourth-order (zero lag) Butterwoth filter using a 2 Hz cut-off frequency.

A strain-gage force platform model OR 6-6 2000 (Advanced Mechanical Technology, Inc., 1991) was used to determine the three components of ground reaction forces, the location, direction and magnitude of external forces. The force platform signals were sampled at 120 Hz and synchronized with the kinematic data using the event and video control unit (Peak Performance Technologies, Inc., 1997).

The body was modeled with five segments (foot, leg, thigh, pelvis, and a segment of head/arms/trunk) with intersegmental joints at the ankle, knee, hip, and L5/S1. Each segment was assumed to have a fixed point mass and constant moment of inertia. Center of mass locations within the segments and the moments of inertia were approximated using anthropometric data and equations obtained from De Leva (1996). The only exception was the pelvis segment. De Leva (1996) defined the lower trunk segment between the omphalion and the hip joint center. The current model utilized a pelvic segment defined by the hip and L5/S1 joints. Therefore the mass of the pelvic segment was estimated to be 11.8% of body mass (Web Associates, 1978). Center of mass and moment of inertia values were modeled as an elliptical solid (Hanavan, 1964) with dimensions determined from anthropometric measurements.

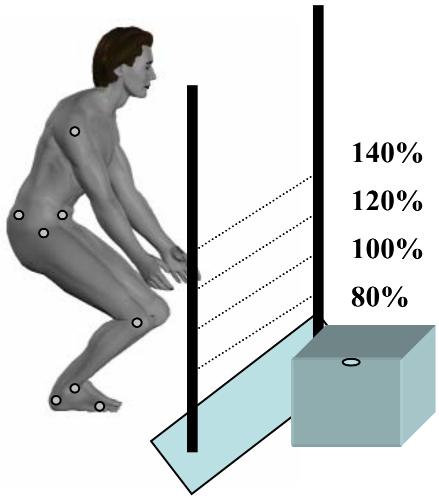

Figure 1 The schematic figure of the three-dimensional positions of nine reflective markers.

The inverse dynamics of the standard link segment model (Winter, 1990) and Newtonian equations of motion starting at the foot were applied to determine the reaction forces and moments at the proximal end of each segment. Equations of motion and the anthropometric model were implemented using a custom analysis program.

To calculate the compressive force on the L5/S1 joint, the following simplifications were used (Chaffin, et al., 1999):

1. No abdominal pressure acted on the diaphragm in front of spinal column (Chaffin, et al., 1999, De Looze et al., 1999),
2. The line of action of the extensor spinae muscles of the lower back was assumed to act parallel to the normal force of compression on the L5/S1 joint and with a moment arm (E) of 6.0 cm (Kumar, 1988),
3. The compression force was assumed to act at the center of rotation of the joint and thus was not considered in the moment equation.

Muscle force (F_M) was then solved as follows:

$$F_M = M_{L5/S1} / E$$

Finally, the forces acting parallel to the disc compression force (F_C) were expressed by

$$F_C = F_M + F_V \text{ cosine } (\alpha) + F_{AP} \text{ cosine } (\alpha)$$

Where, α was the sacral angle, F_V was the vertical reaction force, and F_{AP} was the anteriopostior reaction force. This method neglects the influence of co-contractions on the resultant compressive force.

4 Data Analysis

The independent variable was the barrier height with 5 levels. The main dependent variable was the peak compressive forces at the L5/S1 joint. The time of the peak compressive forces was observed slightly after the time that subjects lift-off the load. In order to determine the cause of differences in the peak compressive forces, several variables were examined at the time of peak compressive forces. These variables included the horizontal distance between the center of load and L5/S1 joint, the vertical acceleration of the load, the angular acceleration of the trunk, the sacral and trunk angles and the postural index. The sacral angle was defined as the angle formed by the base of sacrum (Chaffin, et al., 1999). It was assumed to be 45° during static trial (Thieme, 1950). The postural index was defined as the ratio of the knee angle to the sum of hip and L5/S1 angles.

Means and standard deviations were calculated for each condition. The repeated measures analysis of variance (ANOVA) was performed to detect the variations in the results so that significant differences (table 1) show real differences in the data. In addition, a polynomial regression model of second order was used to analyze the trend in the peak compressive forces after adjusting for the unequal intervals between conditions. An alpha level of 0.05 was selected as the level of statistical significance.

5 Results

Mean (average), minimum (min) and maximum (max) values of the peak compressive forces at L5/S1 for all conditions showed in the Figure 2. It was the compressive forces that are thought to influence the potential for injury.

The other variables are used to explain why the compressive forces changed. There was a statistically significant increased in peak compressive forces at L5/S1 between the non-constrained and the constrained conditions ($p < 0.05$).

Compared to the 0% barrier height the peak compressive forces at L5/S1 were increased by 306.2 N (7.9%), 293.6 N (7.6%), 267.2 N (6.9%) and 210.6 N (5.4%) for the 80%, 100%, 120% and 140% barrier heights respectively (table 1).

	Peak Compressive Forces (N)				
Conditions	Average	Std. Dev	Min	Max	Significance
1 (0%)	3868.8	527.5	3341.3	4396.3	1 vs 2,3,4,5
2 (80%)	4175.0	486.0	3689.0	4661.0	
3 (100%)	4162.4	462.3	3700.1	4624.7	
4 (120%)	4136.0	553.1	3582.9	4689.1	
5 (140%)	4079.4	468.9	3610.5	4548.3	

	Postural Index				
Conditions	Average	Std. Dev	Min	Max	Significance
1 (0%)	0.47	0.19	0.28	0.66	1 vs 2,3,4,5 2 vs 5
2 (80%)	0.40	0.16	0.24	0.56	
3 (100%)	0.38	0.13	0.25	0.51	
4 (120%)	0.39	0.13	0.26	0.52	
5 (140%)	0.36	0.13	0.23	0.49	

	Horizontal Distance L5/S1- Load (m)				
Conditions	Average	Std. Dev	Min	Max	Significance
1 (0%)	0.75	0.38	0.37	1.13	1 vs 2,3,4,5
2 (80%)	0.78	0.38	0.40	1.16	
3 (100%)	0.79	0.44	0.35	1.23	
4 (120%)	0.78	0.38	0.40	1.16	
5 (140%)	0.77	0.32	0.45	1.09	

	Load Vertical Acceleration ($m.s^{-2}$)				
Conditions	Average	Std. Dev	Min	Max	Significance
1 (0%)	3.4	0.90	2.5	4.3	None
2 (80%)	3.6	0.70	2.9	4.3	
3 (100%)	3.5	0.50	3.0	4.0	
4 (120%)	3.7	0.90	2.8	4.6	
5 (140%)	3.4	0.60	2.8	4.0	

	Sacral Angle (degree)				
Conditions	Average	Std. Dev	Min	Max	Significance
1 (0%)	62.4	13.3	49.1	75.7	1 vs 2,3,5 5 vs 2, 3
2 (80%)	66.1	10.4	55.7	76.5	
3 (100%)	67.6	9.8	57.8	77.4	
4 (120%)	68.3	10.4	57.9	78.7	
5 (140%)	70.6	9.5	61.1	80.1	

	Trunk Angular Acceleration ($degree.s^{-2}$)				
Conditions	Average	Std. Dev	Min	Max	Significance
1 (0%)	245.7	111.6	134.1	357.3	None
2 (80%)	275.7	77.2	198.5	352.9	
3 (100%)	274.2	80.9	193.3	355.1	
4 (120%)	260.9	97.4	163.5	358.3	
5 (140%)	252.3	83.8	168.5	336.1	

	Knee Angle (degree)				
Conditions	Average	Std. Dev	Min	Max	Significance
1 (0%)	54.3	8.4	45.9	62.7	1 vs 2,3,4,5 2 vs 5
2 (80%)	50.9	6.2	44.7	57.1	
3 (100%)	48.0	5.1	42.9	53.1	
4 (120%)	48.6	5.9	42.7	54.5	
5 (140%)	43.9	4.8	39.1	48.7	

	Trunk Angle (degree)				
Conditions	Average	Std. Dev	Min	Max	Significance
1 (0%)	3.10	12.3	-9.2	15.4	None
2 (80%)	0.20	8.5	-8.3	8.7	
3 (100%)	-0.20	6.0	-6.2	5.8	
4 (120%)	1.10	7.6	-6.5	8.7	
5 (140%)	0.50	5.7	-5.2	6.2	

Table 1 Means (average), standard deviations, minimum (min) and maximum (max) values of the peak compressive forces at L5/S1 and other variables from all conditions. All variables were measured at the time of the peak compressive forces.

The analysis of the trend of the peak compressive forces at L5/S1 showed statistically significant quadratic trend ($p = 0.0006$). However, the magnitude of the difference of the peak compressive forces during the constrained conditions was small. The peak compressive forces difference between the lowest (80%) and the highest (140%) barrier heights was less than 100 N.

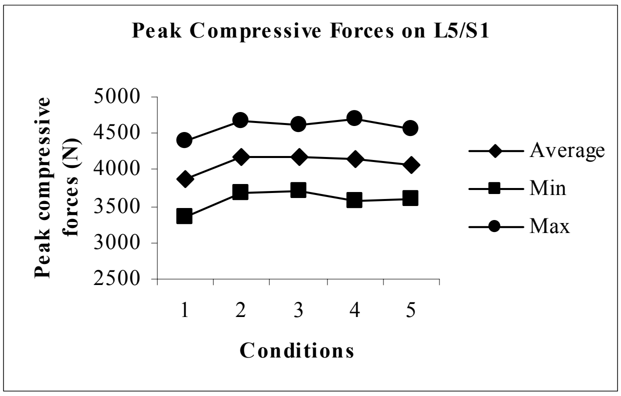

Figure 2 Peak Compressive Forces at the L5/S1 joint for all conditions.

There were significant kinematic changes that occurred, mostly between the 0% barrier height and the other barrier heights (table 1).

The load was moved further from L5/S1 when the barrier was introduced. Compared to the 0% barrier height the load was moved further from L5/S1 in the horizontal direction by 3 cm, 4 cm, 3 cm and 2 cm for the 80%, 100%, 120% and 140% barrier heights respectively (figure 3).

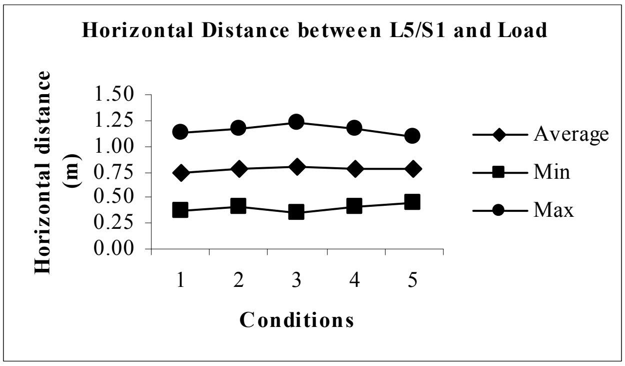

Figure 3 Horizontal Distance between L5/S1 joint to the load for all conditions.

The forward tilt of the pelvis increased as the barrier height increased. This caused an increase in the sacral angle of 3.7 degrees, 5.2 degrees, 5.9 degrees and 8.2 degrees for the 80%, 100%, 120% and 140% barrier heights respectively (figure 4).

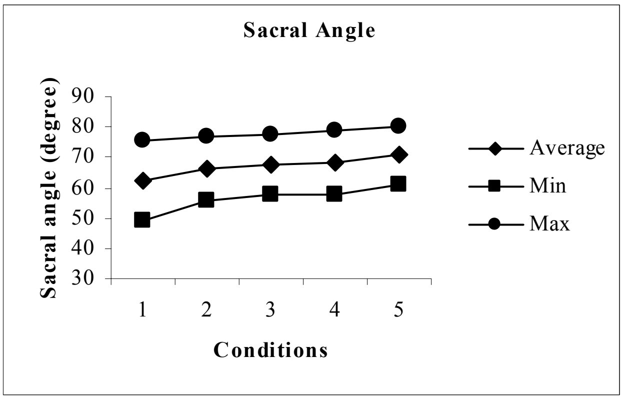

Figure 4 Sacral Angle for all conditions.

There was a trend to decrease the postural index as the barrier height increased. Compared to the 0% barrier height the postural index decreased by 0.07, 0.09, 0.08 and 0.11 for the 80%, 100%, 120% and 140% barrier heights respectively (figure 5).

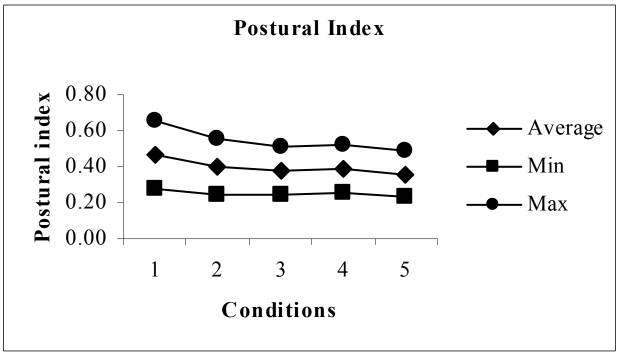

Figure 5 Postural Index for all conditions.

The decrease in the postural index was mainly due to the reduction of the knee angle (figure 6).

The trunk angle, the angular acceleration of the trunk and the vertical acceleration of the load showed no statistically significant changes.

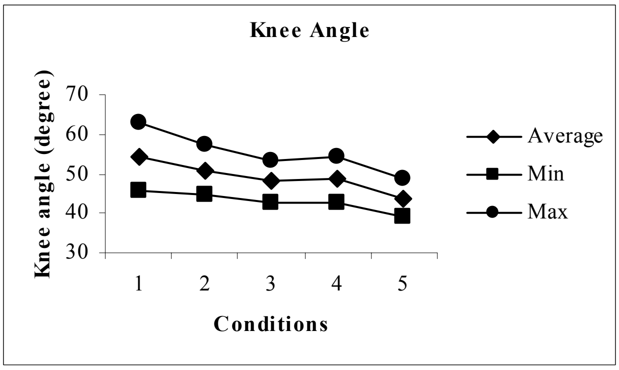

Figure 6 Knee Angle for all conditions.

6 Discussion

The current study applied a dynamic biomechanical model to estimate the mechanical stresses and kinematic changes that occurred as a result of lifting a load over a constraining barrier. Peak compressive forces and moments were consistent with previous studies. De Looze et al. (1992) reported peak moment values at L5/S1 ranged from 220.1 ± 25.2 N•m for an 18.8 kg load. In the present study, moment values at L5/S1 ranged from 215.9 ± 9.4 N•m while lifting a 10.3 kg load during the non-constrained condition. Chen (2000) focused on the peak compressive forces at L5/S1 and found that lifting a 10 kg load produced an average value of 3690 ± 41 N for subjects with an average mass of 67 kg. The mean peak compressive forces at L5/S1 for subjects in the current study (average mass: 75 kg) was 3869 ± 528 N while lifting a 10.3 kg load during the non-constrained condition.

Subjects significantly increased peak compressive forces at L5/S1 when they had to lift the load over a barrier. This was because subjects moved the load further from L5/S1 in the horizontal direction when lifting during the constrained conditions. The greater displacement of the load relative to lower back generates the larger stresses at the lower back. This result supports the previous study (McKean and Potvin, 2001), which showed that lifting over the constraining barrier produced larger horizontal distances from hands to ankles than lifting under the freestyle (no barrier) condition. The larger distance contributes to increased the extensor moment demands and this was confirmed by the increased erector spinae muscle activity.

Increasing barrier height resulted in an increase of the sacral angle and decrease of the postural index. The increase in the sacral angle was due to an increase in forward pelvic tilt. The decrease in the postural index was mainly due to the reduction in the knee angle. These results agree with McKean and Potvin (2001) who found lifting over the constrained condition significantly reduced the amount of flexion at the knees and increased the peak trunk flexion compared to the freestyle condition. Potvin et al. (1991) and McKean and Potvin (2001) discovered the main contributor to increases in the peak trunk flexion would come from the pelvic flexion.

There was a significant quadratic trend in the peak compressive forces. Subjects tended to produce greater peak compressive forces at L5/S1 when lifting over the constraining barriers, where the height was close to knee height (80% - 100%) compared to the constraining barriers, where the height was higher than knee height (120% - 140%) or lower than knee height (0%). This situation prevented subjects from bringing the load closer to the body and thus greater peak compressive forces were generated in the lower back. In addition, the trajectory of the load needs to be more vertical when the load must be lifted over the barrier. When lifting over the higher barrier than knee height, subjects increased the forward pelvic tilt and might move the load closer to the body. An increase of the forward pelvic tilt means a greater sacral angle. The greater sacral angle does not decrease the spinal loading, but it does a shift the loading from compressive to shear.

7 Conclusions

In the current study, when lifting over a constraint barrier, individuals significantly increased the peak compressive forces in the lower back. This was because the constrained condition produced greater horizontal displacement of the load to the lower back than did the non-constrained. Lifting load over a constraint barrier also generated an increase of sacral angle and a decrease of the knee flexion.

The peak compressive forces in the lower back in this study showed a quadratic trend. Therefore, individuals tended to produce greater stresses in the lower back while lifting a load over constraining barrier where the height was close to knee height compared to a constraining barrier where the height was higher or lower than knee height. This fact seems to be a threshold value at which higher heights do not cause increased compressive forces is unique. However, the magnitude of the difference of the peak compressive forces was very small. Thus, a future research should investigate the lifting tasks over constraining barrier height between 0% and 80% and above 140% of knee height.

To prevent the risks of the lower back injury, people should avoid lifting load over a barrier if possible. This is because the constrained condition significantly generates a larger horizontal displacement between the load and the lower back, which will produce greater stresses on the lower back. So, when lifting people should keep the load close to the body. Lifting load under the constrained condition requires that people use more back than knees. This situation also develops more stresses on the lower back. If a barrier is necessary, it may be marginally beneficial to avoid barrier heights that restrict knee flexion.

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